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AUSTRALIA

Patents Act 1990

University of Technology, Sydney

PROVISIONAL SPECIFICATION

Invention Title:

Microwave Antenna for Cardiac Ablation

The invention is described in the following statement:

Technical Field

The invention concerns a microwave antenna for cardiac ablation. The invention further concerns a cardiac ablation system and method.

Background Art

The heart is composed of three types of cardiac tissue. They are atrial muscle, ventricular muscle and specialized excitatory and conduction tissues. The atrial and ventricular muscles of the heart are normally excited synchronously. Each cardiac cycle begins with the generation of action potentials by the sino-atrial, SA node located in the posterior wall of the right atrium. These action potentials spread through the atrial muscle by means of specialized conduction tissue, causing contraction. The action potentials do not normally spread directly from the atrial muscles to the ventricular muscle. Instead, the action potentials conducted in the atrial musculature reach the atrioventricular, AV node and its associated fibers, which receive and delay the impulses. Potentials from the AV node are conducted to the His-Purkinje, HIS bundle. This structure carries the impulses to the ventricular musculature to cause their synchronous contraction following contraction of the atrial muscles. Arrhythmia is a term used to describe irregular beating of the heart.

Episodes of an abnormal increase in heart rate may occur, and are termed paroxysmal tachycardia. This can result from an irritable focus in the atrium, the AV node, the HIS bundle, or in the ventricles. Episodes of tachycardia may be initiated and sustained by either a re-entrant mechanism, or may be caused by repetitive firing of an isolated focus. While these episodes of tachycardia can be responsive to treatment by medication, under many circumstances surgical ablation of the abnormal conducting tissue may be the only cure.

Atrial fibrillation is one of the most common arrhythmias with high mortality rate. Cardiac arrhythmias are generally circuits, which form within the chambers of the heart. Arrhythmia circuits are abnormal electrical connections, which may form in various areas of the heart. For example, arrhythmia circuits may form around the veins and/or arteries. Atrial fibrillation occurs in the atrial of the heart, and more specifically at the region where pulmonary veins are located. Due to the fact that these arrhythmia circuits prevent the heart from beating normally, cutting the arrhythmia circuits is necessary to restore a normal heart beats. Many different cutting patterns may be implemented to cut arrhythmia circuits and, specifically, arrhythmia circuits formed within the atria.

Each year around the globe, millions of people are affected by arrhythmias, most of which can be treated and are not life-threatening. However, many arrhythmias' can be very dangerous and claim about 500,000 lives in the United State of America each year. In Australia, atrial fibrillation, occurs in 0.4% of general population. In the elderly population – those over the age of 80, it has a prevalence of around 10%. One third of all patients who have strokes are in atrial fibrillation when they get to hospital. The atrial fibrillation causes clots in the atria, which travel to the brain to cause the stroke. If a patient goes into atrial fibrillation after a heart attack, the likelihood of fatality doubles.

Of all patients with atrial fibrillation the risk of a stroke or similar problem is around 5% per year if not treated. Current medical treatments rely on pharmaceutical medications and at best the success rate is 50%. Furthermore, these patients may suffer adverse and sometimes life threatening side effects as a result of the medication.

Catheter ablation has recently become an important therapy for the treatment of cardiac arrhythmia's, cardiac diarrhythmias and tachycardia. Cardiac ablation refers generally to creating a lesion, in terms of heat, in the myocardial tissue that results in a cessation of biological functioning of the remaining living or diseased cells in the scar tissue that disrupt normal cardiac rhythms. Prior to the ablation procedure, the electrical activation sequence of the heart to locate the arrhythmogenic sites or accessory pathways are mapped. The mapping helps cardiologists determine the precise location of the above mentioned sites.

An ablation approach which has been used is the use of high voltage, direct current defibrillator discharges. However, direct current ablation has several drawbacks including the need for general anaesthesia and explosive discharges that can rupture certain cardiac tissues. For these reasons, direct current ablation has been rendered obsolete.

More frequently ablation catheter systems used are which utilise radio frequency, RF energy in the hundred kilohertz range as the ablating energy source. Accordingly, a variety of RF based catheters and power supplies are currently available to the electrophysiologist. However, RF energy has several limitations including the rapid dissipation of energy in surface tissues resulting in shallow lesions and failure to access deeper arrhythmic tissue. Another limitation of RF ablation catheter is the risk of clot formation on the energy emitting electrodes. Such clots have an associated danger of causing potentially lethal strokes in the event that a clot is dislodged from the catheter.

Recent advances in catheter ablation technologies have investigated the use of other energy mode delivery systems. Researchers have investigated the use of energy

in optical and ultrasound frequency as ablation energy. Several drawbacks and limitations of these energy sources have been discovered which limit their success.

Microwave frequency energy has long been recognised as an effective energy source for radiative heating of biological tissues and has been used in applications such hypothermia for cancer treatment. Accordingly, in view of the drawbacks of the existing catheter ablation techniques, there has recently been a great deal of interest in using microwave energy as an ablation energy source. At microwave frequencies, energy can radiate from the antenna causing water molecules within the myocardium and other dipoles to oscillate with the applied frequency, thereby resulting in dielectric heating. This then generates friction between the molecules. One of the many advantages of tissue ablation using energy at microwave frequency is that the microwave energy can be delivered to the myocardium without physical contact between the antenna and the myocardium. Furthermore, the energy coupled into the myocardium is impeded by the rise of impedance during ablation procedure. The attainable local tissue temperature ultimately governs the generated lesion size. Some idea of tissue temperature can be obtained by monitoring the temperature surrounding the area near the antenna.

From Table I, we can see that for the treatment of Atrial Fibrillation, where long and linear lesions are required, microwave energy seems to be a more suitable energy source.

	Radio Frequency	Microwave	Laser
Physical contact	Extremely important	Not very important	Extremely important
Lesion Shape	Spot, isolated	Spot or Linear, depends on Antenna design	Spot, isolated
Power delivery control	Easy	Require some Hardware control	Extremely difficult
Lesion Size	Shallow, Wide	Deep, Linear	Hard to control
Continuous lesion	Extremely difficult	Easy	Extremely difficult
Equipment Cost	Relatively inexpensive	Relatively inexpensive	Expensive

Table 1: Comparison between microwave energy source and existing energy sources

slots between the rings advantageously determines the shape of the near-field distribution. The terminal device may be positioned over the exposed end and short circuited to the inner conductor of the antenna element. The size, diameter, dielectric constant and length of terminal device advantageously minimizes the return loss of the antenna which is indicative of an improved energy delivery into the myocardium. The antenna element may further comprise dielectric and metallic rings.

Alternatively, the antenna element may be configured by bending the distal end of the transmission line into an open loop. The open loop may be oriented such that it extends transverse to the longitudinal axis of the transmission line. In such a configuration, the positioning of the antenna within a vein allows the blood to flow through the loop. This configuration has a resultant cooling effect while the ablation takes place.

The antenna may be further configured such that the energy emitted from the antenna is directed substantially transverse to the longitudinal axis of the antenna.

The depth of the lesion formed may be substantially as wide as it is deep.

The antenna may further comprise an integrated temperature sensor to sense the temperature of the tissue being ablated by the antenna.

The antenna may further comprise a Teflon sheath surrounding at least the active antenna section. This advantageously ensures safety and biocompatibility.

The antenna may further comprise an elongated flexible tube for insertion into a vessel in the body of a patient wherein the transmission line is disposed at least partly within the flexible tube.

The frequency emitted from the antenna may have a bandwidth suitable for varying field emission characteristics depending on the lesion required.

The impedance of the antenna may be matched at a frequency of between 900 MHz and 3000 MHz, and more preferably at a frequency of about 2450 MHz.

Advantageously, the microwave antenna is miniaturised which minimises tissue damage during surgical invasion.

In use, the position of the antenna may be adjusted to bring the antenna element into contact with an arrhythmogenic site, or may be positioned so that the antenna element is proximate to an arrhythmogenic site. Sufficient electromagnetic energy is applied to cause thermal damage to the tissues surrounding the antenna. This may be achieved using fluoroscopy.

Many of the existing microwave ablation catheter antennas produce an end fire or forward firing near-field pattern. However, using these antenna catheters is extremely difficult in treating atrial arrhythmia's where linear lesions are required. Linear lesions are important in treating atrial arrhythmia because isolation is required between two conducting nodes. If a linear lesion is not formed, or if there are discontinuities within the lesion, the electrical signal in the heart can actually find another path. This will cause the arrhythmia to re-occur after the procedure. Furthermore, the dimensions of such catheters through which antenna pass are dictated by their requirement to be able to be inserted into veins for the passage to the human heart. Typically, a 6F catheter which has a diameter of approximately 3mm is utilised.

Disclosure of Invention

In a first aspect, the invention is a microwave antenna for cardiac ablation comprising:

a transmission line having an inner conductor, an outer conductor and a dielectric insulator to provide insulation between the inner and outer conductor; and

an energy emitting antenna element at the distal end of the transmission line for generating a microwave near-field to cause localised tissue heating, the antenna element comprising an inner conductor an outer conductor and a dielectric insulator coupled to the respective inner and outer conductors and insulator of the transmission line, the antenna element further comprising a terminal device at its distal end; wherein the antenna element is configured such that in use the microwave energy is focussed to form a substantially linear lesion.

The coupling of the antenna element to the transmission line may be such that the antenna element is formed in the distal end of the transmission line. Such integration would ensure high physical integrity.

The transmission line is selected for its suitability for insertion into veins of a human body, in order to position the antenna element in organs such as the heart. Characteristics which determine a transmission lines suitability may include its flexibility, strength, diameter, and operating temperature range. For example, the transmission line may be a commercially available cable such as those traded under the name TFlex®.

In one embodiment, the antenna element may be configured such that a portion of the outer conductor of the transmission line is removed and a plurality of conducting rings are spaced along the exposed insulator. The size and positioning of the rings and

In a second aspect, the invention is a cardiac ablation system comprising:

a generator to generate microwave energy;

a transmission line in communication with the microwave generator, the transmission line having an inner conductor, an outer conductor and a dielectric insulator to provide insulation between the inner and outer conductor;

an energy emitting antenna element at the distal end of the transmission line for generating a microwave near-field to cause localised tissue heating, the antenna element comprising an inner conductor an outer conductor and a dielectric insulator coupled to the respective inner and outer conductors and insulator of the transmission line, the antenna element further comprising a terminal device at its distal end, wherein the antenna element is configured such that in use the microwave energy is focussed to form a substantially linear lesion;

a temperature sensing element integrated with the antenna; and

control logic in communication with the microwave generator to receive input conditions and to control the therapeutic characteristics of the lesion based upon the sensed tissue temperature and the input conditions.

Input conditions may include, but not be limited to: setting the maximum output power of the generator, the ablation duration, the upper threshold tissue temperature and the lower threshold tissue temperature.

The control logic may control the power output of the generator and may control the length of the ablation procedure.

The therapeutic characteristics may include but not be limited to the size and shape of the lesion, and whether the lesion is continuous or non-continuous.

Advantageously, the microwave antenna may minimise the risk of the antenna element separating from the transmission line when the antenna is inserted into or removed from a patient's body.

Advantageously, the cardiac ablation system offers an alternative treatment modality for patients whose heart rhythm disorders are not responsive to drug therapy and are too weak to go through open-heart surgery. The system advantageously offers the treatment efficacy of the open-heart surgery without the trauma and post-operative intensive care associated with the open-heart surgery.

The antenna may appeal to hospitals worldwide that perform cardiac catheter ablation. Furthermore, the system can be integrated quite readily into hospitals throughout the world that performs cathetization and electrophysiology studies.

Brief Description of Drawings

An example of the invention will now be described with reference to the accompanying drawings, in which:

Fig. 1 is a cross sectional schematic of a coaxial slot array antenna;

Fig. 2 is a graph illustrating the return loss of the antenna shown in figure 1;

Fig. 3 is a graph showing the normalised specific absorption rate SAR of tissue under the irradiation of the antenna shown in figure 1;

Fig. 4 is a graph illustrating the normalised SAR distribution of the antenna shown in figure 1;

Fig. 5a is a top view of an parallel loop microwave antenna;

Fig. 5b is a left elevation view of the parallel loop microwave antenna;

Fib. 5c is a right elevation view of the parallel loop microwave antenna;

Fig. 5d is a perspective view of the parallel loop antenna;

Fig: 6 is a graph illustrating the measured and predicted return loss of the PL Antenna;

Fig. 7 is a schematic illustration of a cardiac ablation system;

Fig. 8 is a schematic illustration of a graphical user interface;

Fig. 9 is a schematic illustration of an experimental set-up for testing of the cardiac ablation system;

Fig. 10 is a circuit diagram for the control logic;

Fig. 11 is a schematic illustration of a mode accessed via the graphical user interface illustrated in fig. 8;

Fig. 12 is a graph illustrating the temperature modulated power delivery mode;

Fig. 13 is a flow chart of the temperature control logic;

Fig. 14 is a graph of the return loss comparison between the CSA antenna and the antenna[1,2];

Fig. 15 is a graph of the temperature comparison between the CSA antenna and antennas[1,3,4].;

Fig. 16 is a bar graph illustrating the lesion size for the prior art spiral antenna[3]; and

Fig. 17 is a bar graph illustrating the lesion size for the CSA antenna in accordance with the invention.

Best Mode for Carrying Out the Invention

Figure 1 illustrates a coaxial slot array CSA antenna 10 for cardiac ablation. The antenna 10 has transmission line 11 and an antenna element 13 formed in the distal end

of the transmission line 11. The transmission line comprises an inner conductor 12, an outer conductor 14 and a Teflon dielectric insulator 16 which provides insulation between the inner 12 and the outer conductor 14 of a coaxial transmission line. The insulator has a diameter of about 3 mm, and the conductors about 0.91 mm. To form the antenna element, at the distal end of the transmission line, the outer conductor 14 is striped off and a series of copper rings 18 are fabricated, pairs of which are respectively separated by a slot. The rings 18 are strategically positioned according to the results of computer simulations. Since the insulation layer between the inner conductor and the copper ring is Teflon, dielectric spacers (not shown) are necessary in order to maintain the spacing between respective metallic rings 18. A termination device 20 terminates the distal end of the array antenna where it is short-circuited to the inner conductor 12 of the transmission line. In this embodiment, the transmission line is a commercially available cable TFlex®-402. The operating temperature range of this cable is from –65 to 125 C and has an impedance of 50 ± 1Ω.

The antenna array 10 radiates electromagnetic energy coherently and confines heat to the immediate area surrounding the slot arrays. In order to find the optimum positions for the ring 18 placements as well as the ring 18 sizes, Finite Difference Time Domain techniques FDTD are employed to model the interaction between the microwave antenna and the heart muscle, the myocardium.

The antenna array 10 operates at a frequency of 2450 MHz. The desired length of the radiating antenna is determined by examining the desired requirements of the lesion. Furthermore, the size of the termination device determines the minimum return loss of the antenna, whilst the size and positioning of the rings and slots determines the shape of the near-electric field distribution. During initial experiments it was observed that for a 20mm long antenna, the 3mm termination device gave the best return loss. However with these measurements, the resonant frequency of the antenna was at around 1000 MHz. In order to shift the resonant frequency up to the desired 2450 MHz the slots were reduced to a diameter of 0.5mm and the rings to a diameter of 1mm. The overall length of the antenna is only 18mm and has an overall diameter of 2.7mm which fits comfortably into a 6F catheter.

Figure 2 illustrates the measured 24 and simulated 26 return loss of the CSA antenna using these dimensions. Using the computer simulation technique, the impedance of the coaxial slot array antenna is able to be matched at any desired frequency between 900 MHz to 3000 MHz.

Figure 3 shows the normalised Specific Absorption Rate SAR of the tissue under the irradiation of the CSA antenna. The SAR gives a measure of the rate at

which the microwave energy is absorbed by the tissue under ablation. It can be seen that most of the heating is confined to the antenna element 27 and furthermore it shows that there is no cable heating which is a critical factor to be considered when designing microwave antennas for catheterisation. Typically, the catheters are inserted into the patient via the veins. Therefore there is a large length of the coaxial transmission cable actually within the patient's vein during the ablation procedure.

As illustrated in figure 4, the active array is between 0mm and 20mm. The highest exposure to the electric field is around the centre of the antenna at about 10mm and therefore the highest SAR distribution. The rate of SAR distribution change between 0mm and 20mm is very smooth. The smooth rate of change of the SAR distribution along the active antenna section is critical in producing long linear lesions. In addition, a smooth rate of change of the SAR distribution means that the tissues surrounding the CSA antenna are evenly heated and therefore hot spots in the tissue are eliminated.

Figures 5a to 5d illustrate various views for a further optional microwave antenna for cardiac ablation, the parallel loop antenna (PLA). This antenna is ideally suitable for the ablation of the pulmonary veins and it produces a circumferential radiation along the active section of the loop. With regard to pulmonary veins, it is important to produce continuous linear lesions in order to electrically isolate it from the atrium and thereby stopping any propagation of the undesirable activation signal from reaching the atria.

The parameters of the antenna element of the PLA are as follows, the outer diameter of the coaxial delivery cable 60, the separation distance between the end of the loading cap to the outer conductor of the delivery coaxial cable 62, the separation distance between the outer diameter of the loading cap to the beginning of the open loop antenna 64, the separation distance between antenna/coax junction and the starting of the loop antenna 66; the insert distance of the Teflon dielectric into the end cap 68, the length of the end cap 70, the length of the loading tip which is equivalent to 68 subtracted from 70, the inside loop radius 74 and the bending radius 76.

Using the FDTD method, results from experiments obtained iteratively the optimum dimensions for the constitutive parameters of the PLA. The measured 78 and simulated 79 return loss for the PLA is shown in Figure 6.

As can be seen, the PL antenna is very well matched at the intended frequency of 2450 GHz. It also has a measured voltage standing wave ratio: VSWR<2 and a bandwidth of approximate 1GHz. This will ensure that the PL antenna's performance is not greatly affected by the changes in the dielectric properties of the pulmonary

veins. This makes the PL Antenna robust in terms of the microwave power delivery. Another characteristic of the PL antenna is its donut shaped active antenna section. The hollowed section of the PL antenna allows the blood to flow while the ablation takes place. This is important as the temperature inside of the loop, as well as outside of the loop, will be raised quite considerable during ablation. The blood flow will ensure that the inside of the PL antenna is cooled as to avoid any damage to the veins.

Figure 7 shows a system 80 for performing ablation on myocardium tissue. The system includes a commercially available microwave generator 82 that delivers microwave energy at 2450 MHz. A steerable catheter 84 carrying one of the previously mentioned microwave antenna elements 86 that has integrated temperature-sensing elements 88 is connected to the microwave generator 82 via the coaxial transmission line. The microwave generator 82 is linked to the controlling desktop computer 90 via a DAQ remote interface 92 developed especially for this system. The data transfer is done via ribbon cables. The DAQ Remote Interface 92 is responsible for acquisition of the temperature as well as sending control information to the microwave generator 82. The DAQ Remote Interface 92 and the program running in the desktop computer 90 form an integral part of the control logic. A schematic front panel of the DAQ Remote Interface 92 is shown in figure 8.

Experimental Testing

Figure 9 shows the experimental setup for testing the microwave cardiac ablation system 80. It comprises of a laptop PC 94, temperature sensor, 96, microwave generator 82, tissue bath 98, roller pump 100 and ambient temperature regulator 102. All this equipment is mounted on the especially designed mobile bench 104. The laptop PC 94 is responsible for performing the control algorithm and the acquisition of the tissue temperature through the DAQ Remote Interface board (not shown). The circuit diagram for the DAQ Remote Interface Board is shown in Figure 10. The tissue bath 98 is filled with blood and the cardiac tissue is held in place with specially designed tissue holding grid (not shown). To simulate the realistic situation, the blood flow of the experiment is done using the roller pump 100, which can be set at various flow rates in terms of litre per minute. The ambient temperature regulator 102 keeps the blood at the simulated body temperature of 37°C.

Figure 11 shows one of the ablation modalities accessible via the graphical user interface where a cardiologist can enter the desired ablation duration 110. It also has a graphical and digital feedback of the tissue temperature 112. The tissue temperature is plotted over the whole ablation period so the characteristics of the tissue temperature

characteristics can be easily conveyed. The software is also equipped with a soft-emergency stop button 114 so the cardiologist can terminate the ablation procedure at anytime. On the other hand, if from the temperature-time curve 112, the cardiologist feels that the ablation duration should be extended, they can simply modify the "OFF-Time" 116, while the ablation procedure is taking place.

Figure 12 shows the power delivery control based on temperature modulation. The switching on and off of microwave energy delivery is modulated by the temperature bounded by the Upper and Lower Threshold Tissue Temperature (UTTT and LTTT respectively).

Figure 13 shows the preferred temperature control logic. In this embodiment, the temperature monitoring process 120 constantly monitors the temperature of the tissue site undergoing ablation based on the desired Upper and Lower Threshold Tissue Temperature (UTTT and LTTT respectively) 122.

The preferred predetermined tissue temperature setting depends upon the desired therapeutic characteristic of the lesion. Typical therapeutic lesion characteristics are the surface area of the tissue that is ablated and the depth of the ablation. The UTTT is limited to 100 degrees C. By operating at the desired tissue temperature condition, the temperature monitoring process 120, controls the therapeutic characteristics of the lesion during the ablation procedure. While the temperature monitoring process 120 controls the power level of the output microwave power, the main timer loop 110 controls the length of the ablation procedure. A database of these values is stored in the criteria database 124 and is associated to the ablation modality chosen 126. However, the operators have the ability to overwrite the maximum power setting 128 and the desired tissue temperature 122.

At the start of the ablation procedure, the microwave generator is switched to the maximum power set by the cardiologist using process 128. Once the temperature has reached the UTTT set in process 122, the control logic will switch off or decrease the output power of the microwave generator. The tissue temperature monitoring process keeps recording the tissue temperature and once the tissue temperature falls to or below the LTTT, the tissue temperature monitoring process 120 then instructs the microwave generator to increase its output power 132. The procedure is then repeated till the main timer 110 ends.

Comparative Evaluation with Existing Technology

The performance of the CSA antenna, in terms of its return loss, SAR and lesion size, is compared with the microwave antennas in the background art literature.

Comparison is made with the metal tip monopole[1], the dielectric-coated disk-loaded monopole including sleeve choke[2], the spiral wide-Aperture Microwave Cardiac Ablation Antenna[3] and the helical antenna[4].

Utilising novel miniaturised applicators in accordance with the invention, microwave energy is delivered to the heart tissue where abnormal heart rhythm occurs. Once the antenna is employed and positioned, microwave energy is delivered which cause tissue necrosis, thus stopping the arrhythmia. While the miniaturised applicator is responsible for delivering microwave energy to the heart, the software constantly monitors the ablation process. With the in-built intelligent monitor, the software constantly monitors the critical ablation parameters in order to ensure patients safety and ablation efficacy.

Antenna Return Loss Comparison:

The comparison of the antenna's return loss is shown in figure 14. As illustrated, the CSA antenna in accordance with the invention has better return loss compared with the antenna[1 and 2]. This means that during the ablation procedure, more power is coupled into the myocardium and hence will have higher efficiency. The CSA antenna not only has high-energy transfer efficiency, due to its low return loss, but the cable heating due to reflected microwave energy is also minimised. This is shown in Figure 3, where there are virtually no SAR values along the transmission cable (from 0mm to 60 mm on the y-axis). It is not possible to compare the spiral wide-aperture antenna[3] and the helical antenna[4] in terms of return loss because they operate at lower frequencies. Using the return loss comparison, we are confident that the CSA antenna in accordance with the invention has better efficiency while minimising cable heating.

Antenna Temperature Profile Comparison:

Figure 15 illustrates the comparison of the temperature obtainable of the CSA antenna and the antenna[1, 3, 4] cited in the references. In order to set a common standard in measuring temperature for the various antennas, the 75 °C was set as upper threshold for the temperature and sixty seconds for the time. The reason for this is that in order to achieve irreversible lesions, temperature greater than 55 °C must be obtained. But the temperature should stay below 90 °C to avoid tissue vaporisation. It can be seen that to achieve similar temperature results, the CSA antenna only requires half (50%) of the input power required by other antennas. Also it was able to reach the upper temperature threshold in less time than the other antennas.

Lesion Size Comparison:

Figure 16 and 17 show the comparison of the lesion width and depth of the spiral antenna[3] and the CSA antenna. In the ablation procedure it is extremely difficult to achieve a 1:1 ratio between width and depth due to the thermal conductivity properties of the cardiac tissue. From the chart, it is evident that the lesions produced by the Spiral antenna[3] produce lesions, which are wider than they are deep. Therefore, in order to achieve a desirable depth, wide area of the already scarce myocardium tissue will be destroyed. On the other hand the width to depth ratio of the lesions produced by the CSA antenna is much closer to 1:1. Therefore the scarcely available myocardium tissue can be preserved.

Of course the invention is not limited to the application to cardiac ablation, the invention could equally well be applied to the circumstance of prostate cancer treatment, endometrial ablation treatment, and treatment for breast cancer, liver cancer and kidney disease.

It will be appreciated by persons skilled in the art that numerous variations and/or modifications may be made to the invention as shown in the specific embodiments without departing from the spirit or scope of the invention as broadly described. The present embodiments are, therefore, to be considered in all respects as illustrative and not restrictive.

Dated this twenty sixth day of March 2003

University of Technology, Sydney
Patent Attorneys for the Applicant:

F B RICE & CO

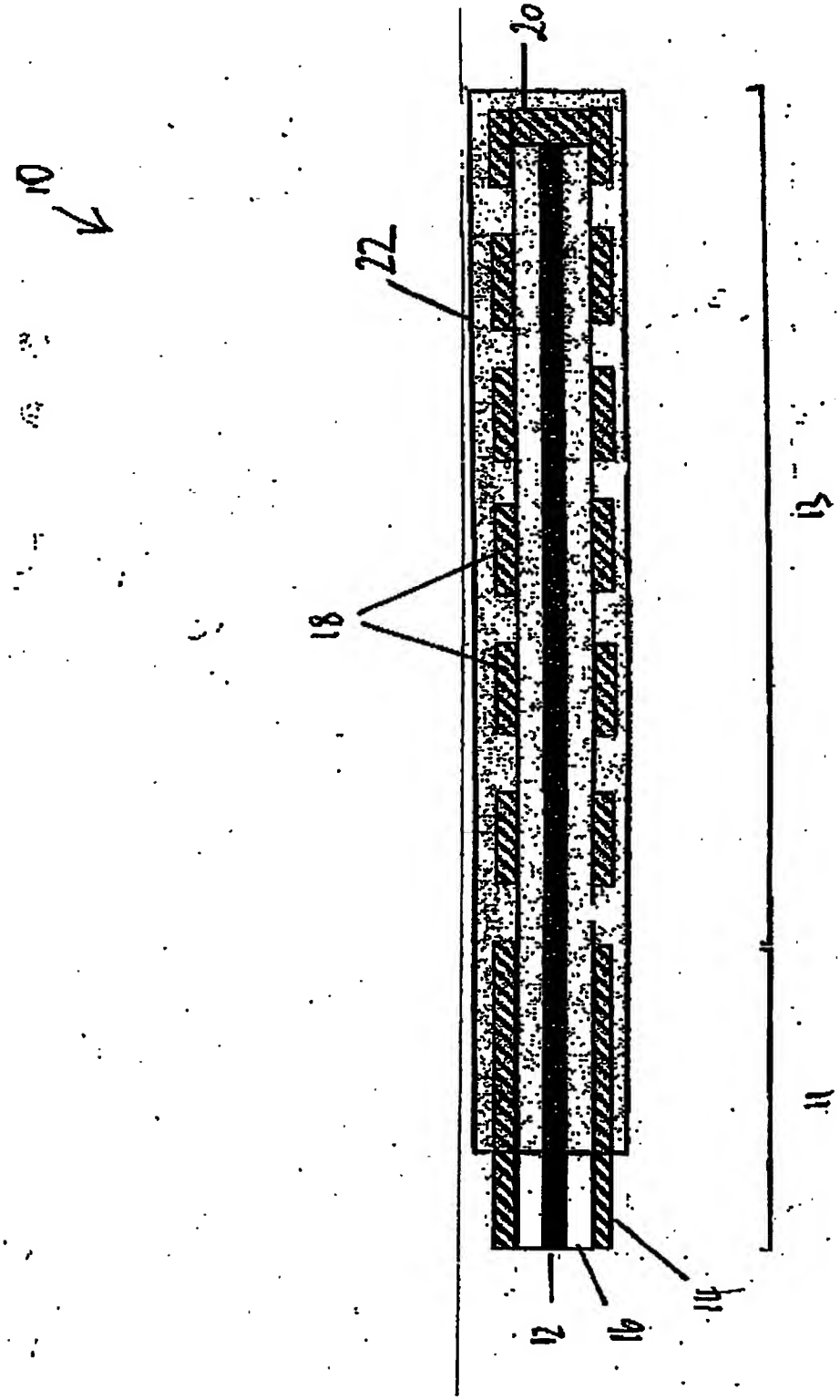

Fig. 1

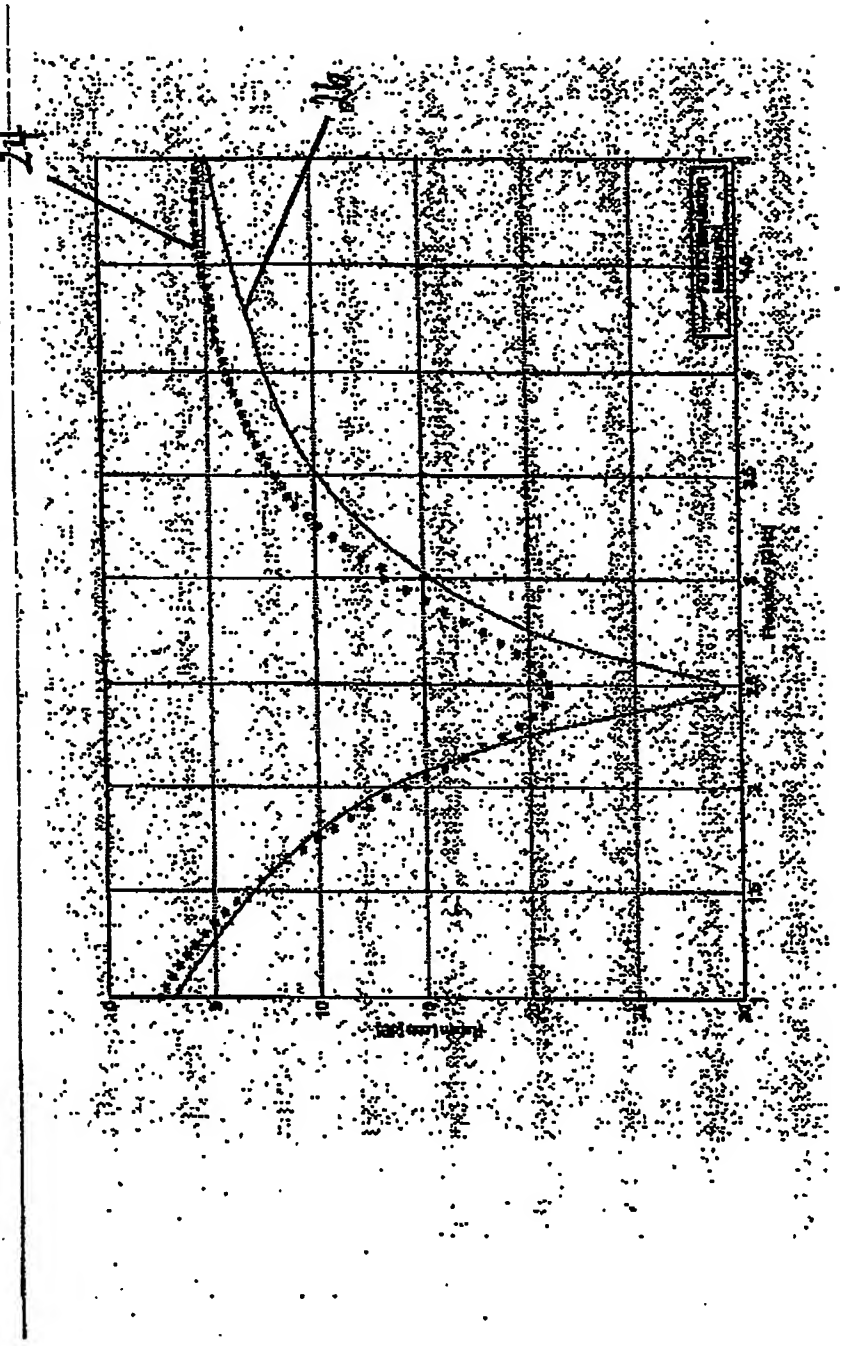

Fig. 2

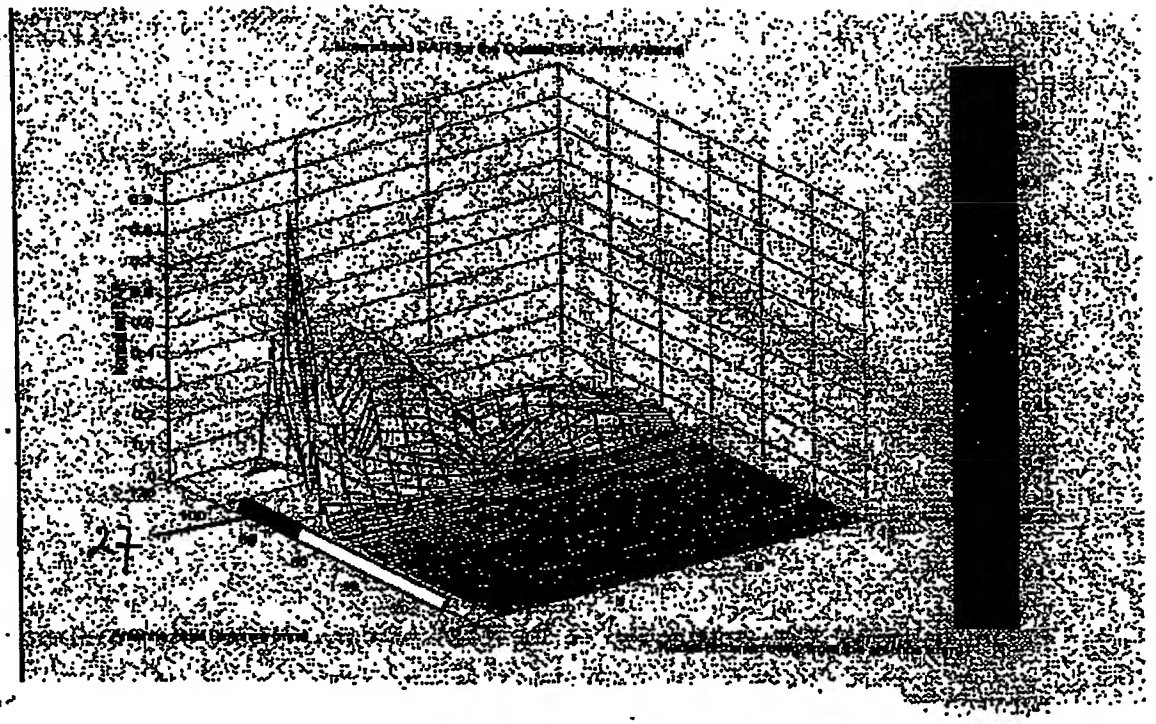

Fig. 3

Fig. 4

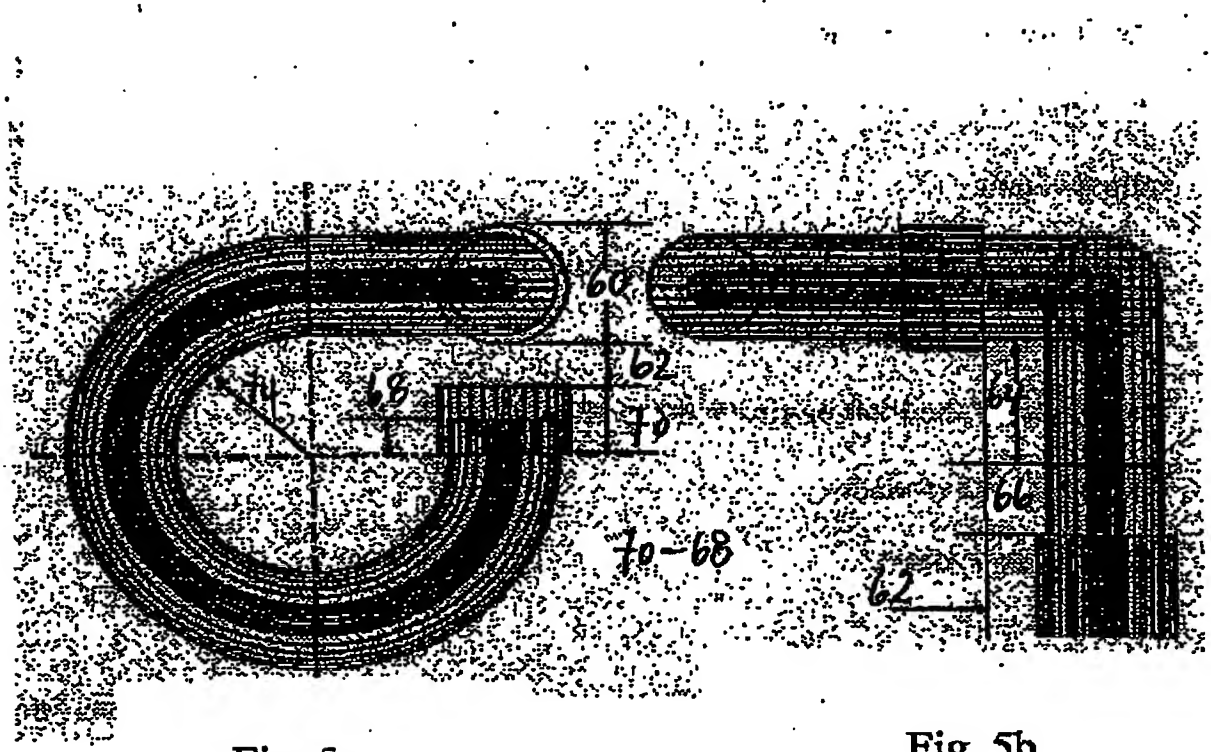

Fig. 5a

Fig. 5b

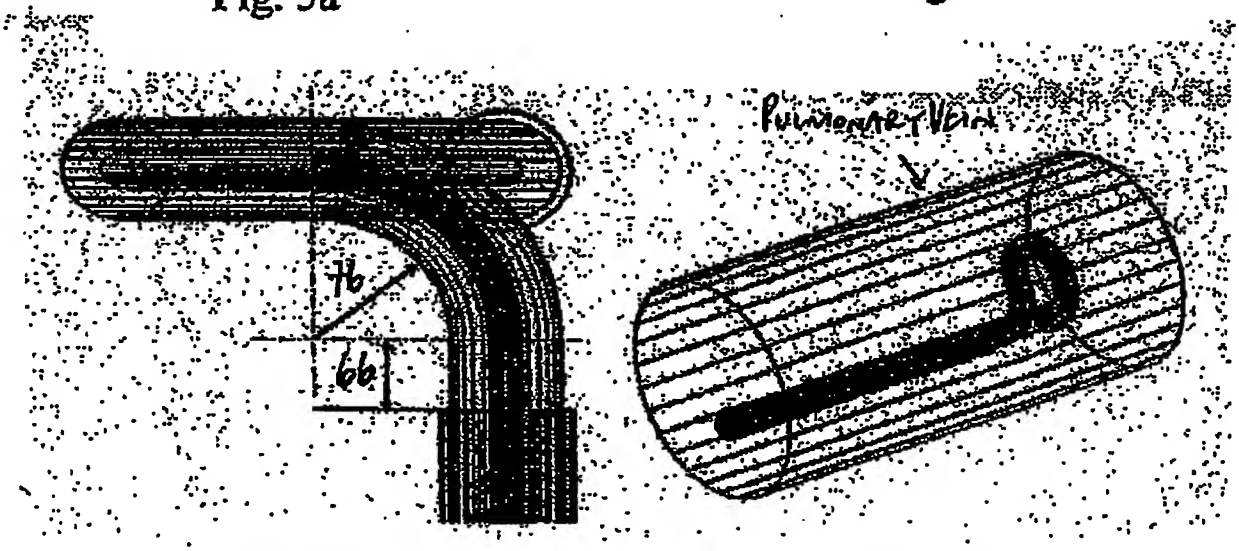

Fig. 5c

Fig. 5d

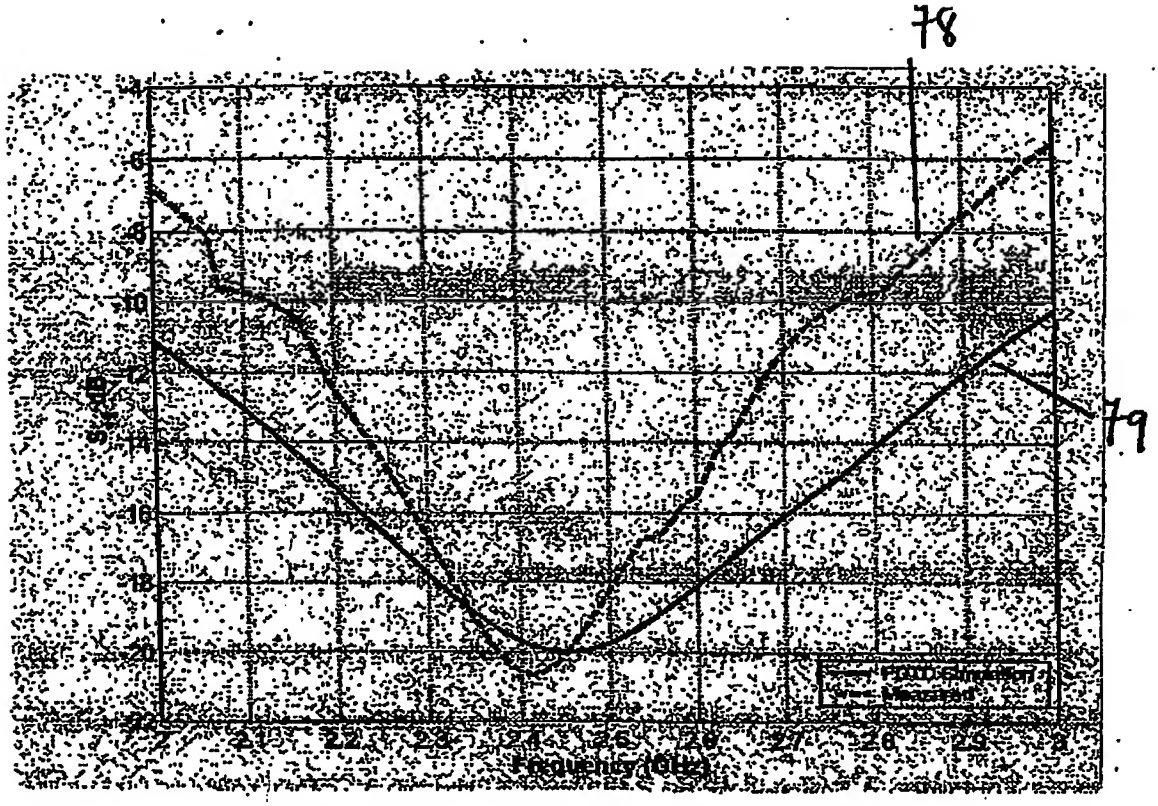

Fig. 6

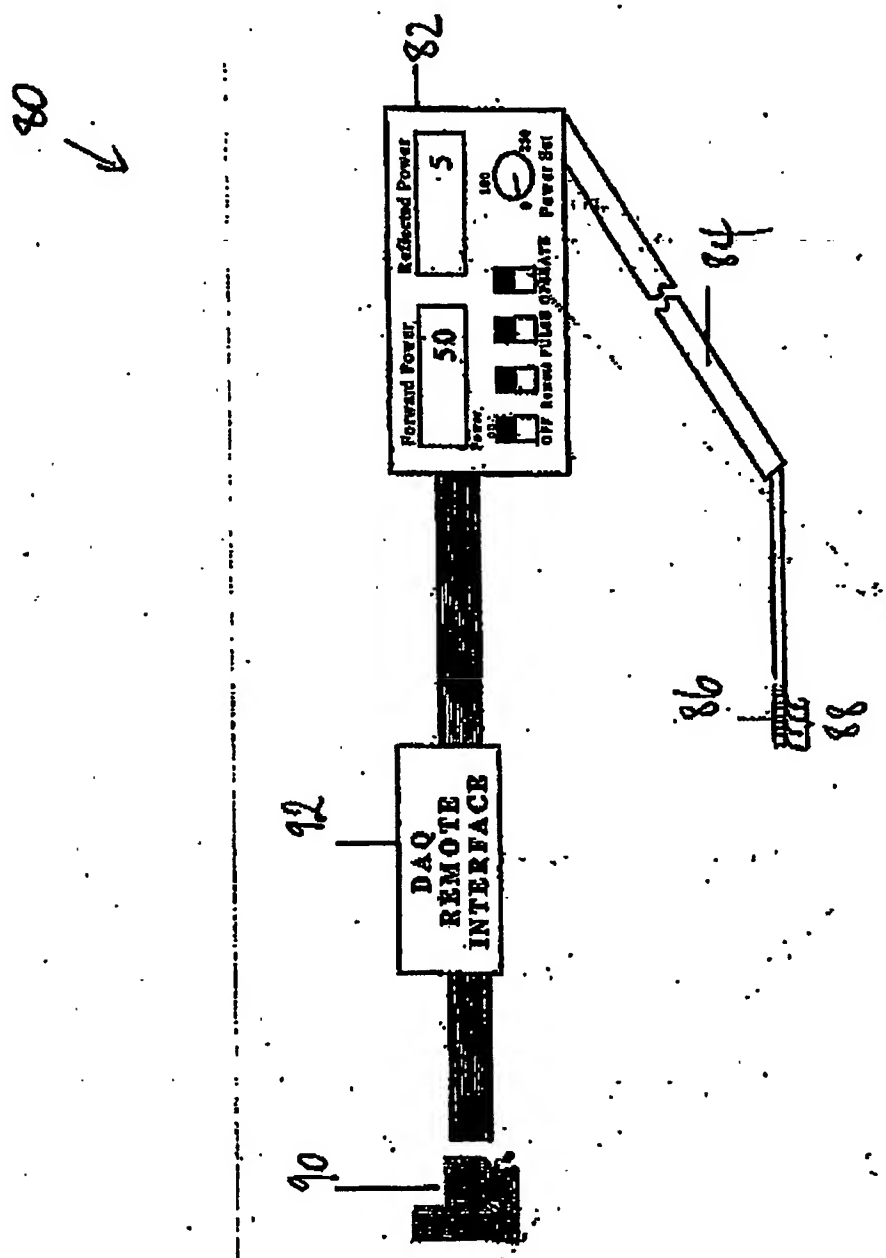
80
82
Reflected Power
5
Power Set
Forward Power
50
84
86
88
DAQ
REMOTE
INTERFACE
12
10

Fig. 7

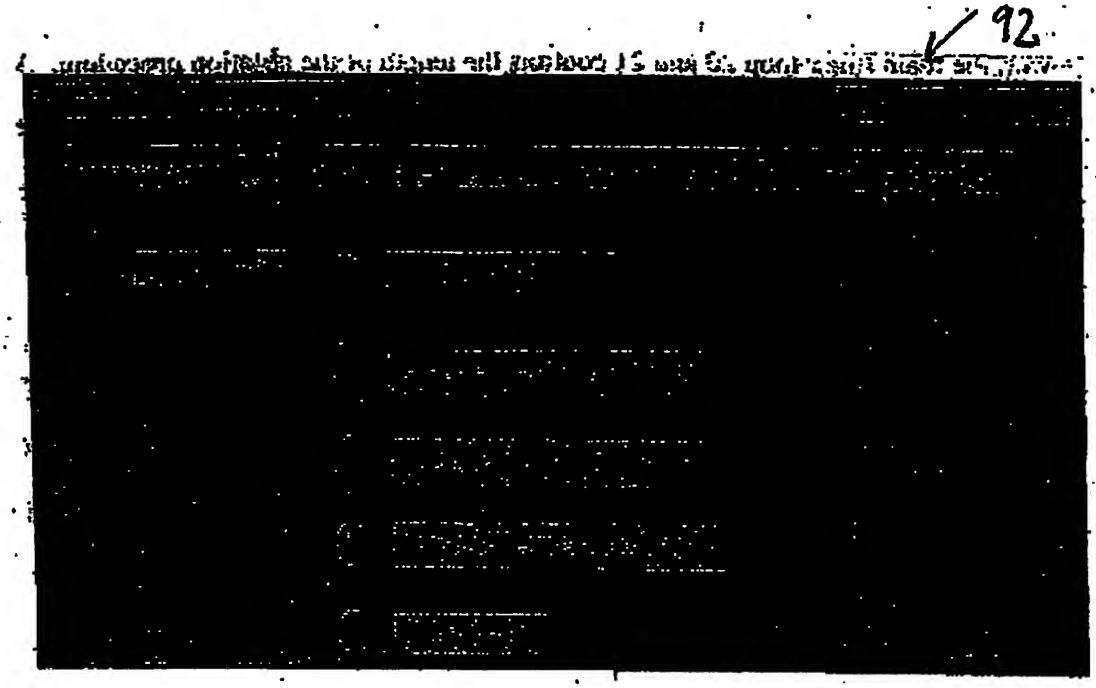

Fig. 8

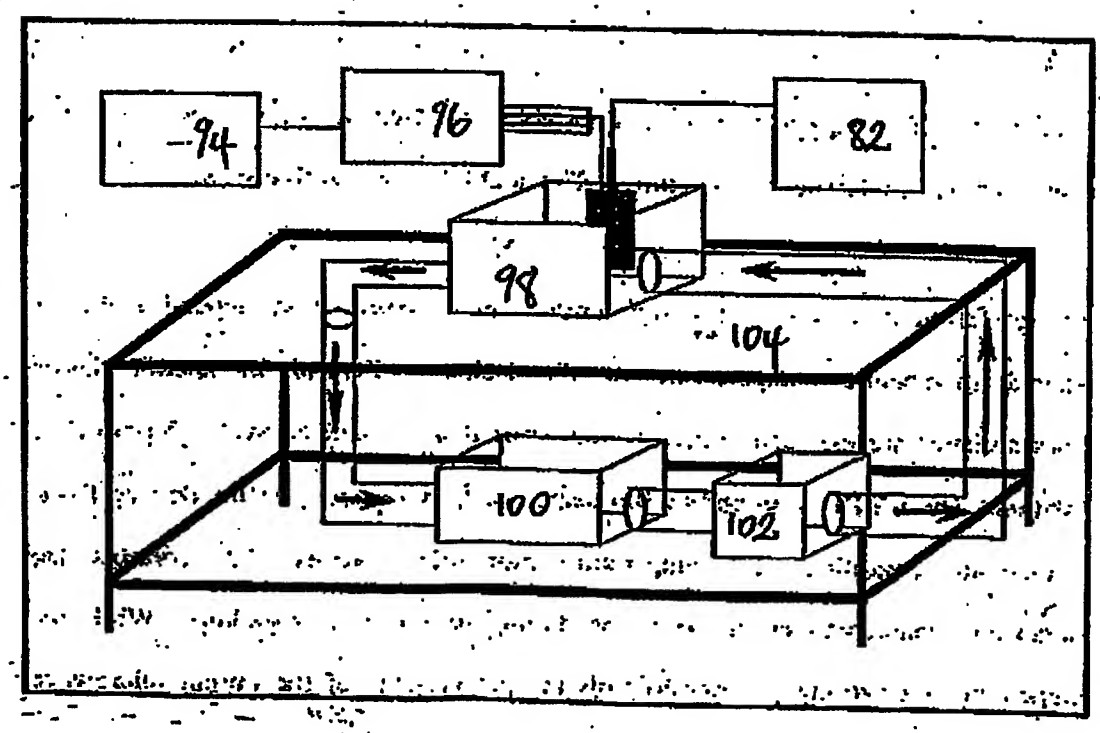

Fig. 9

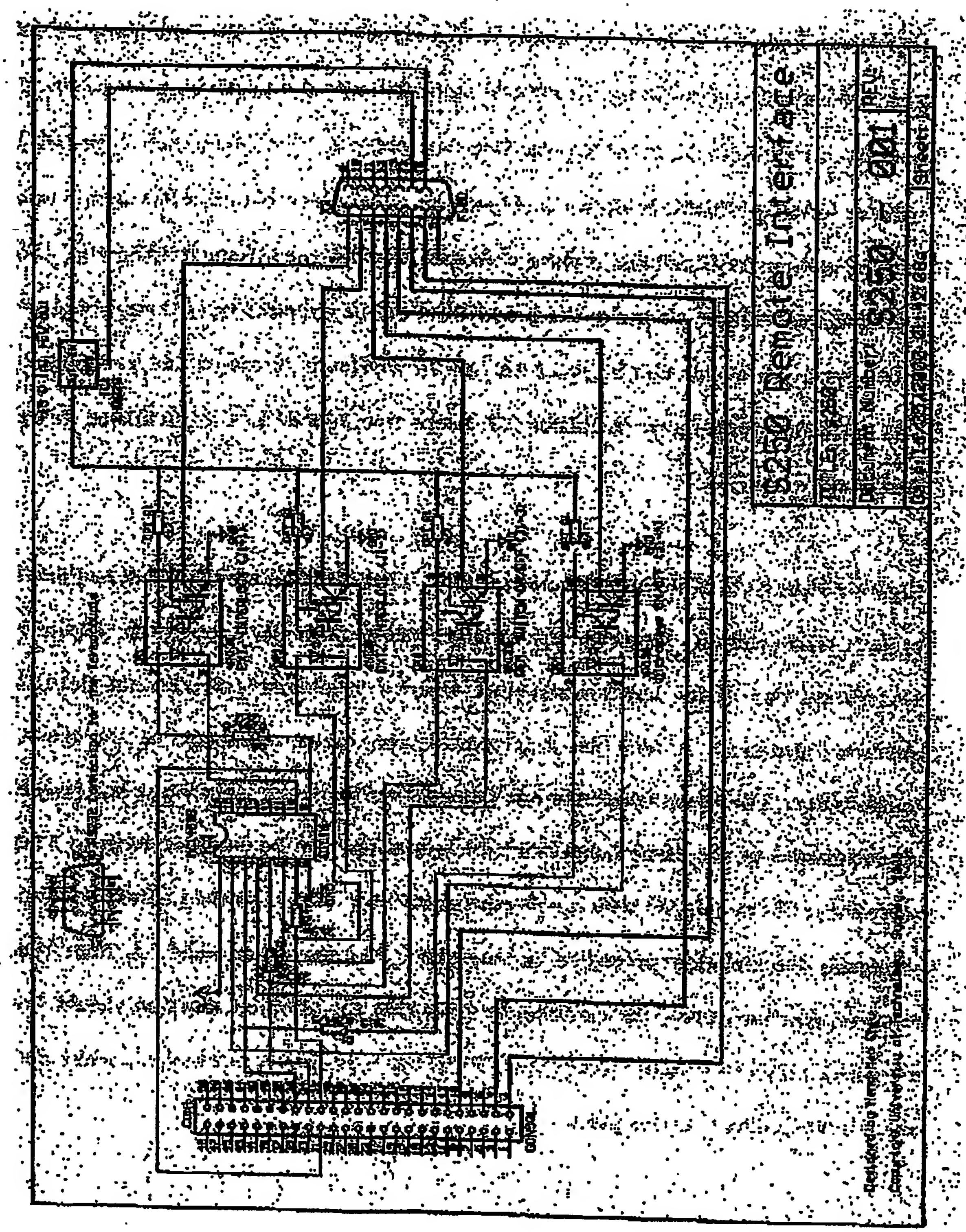

Fig. 10

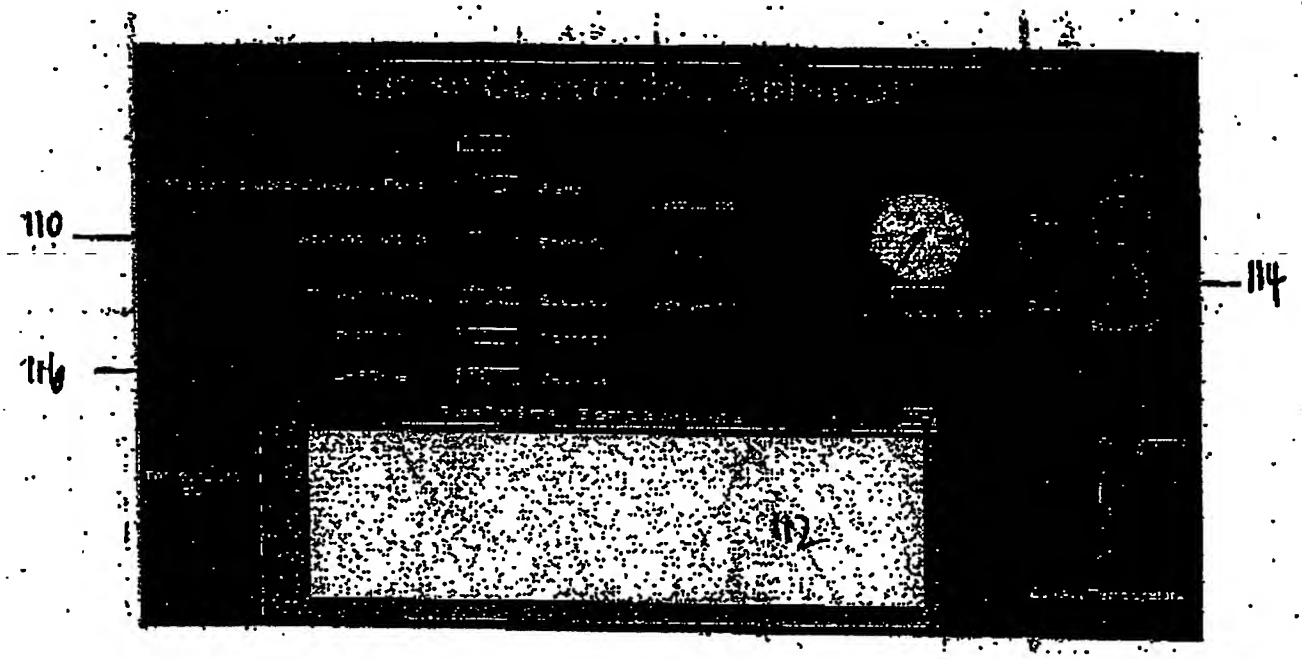
110
114
116
112

Fig. 11

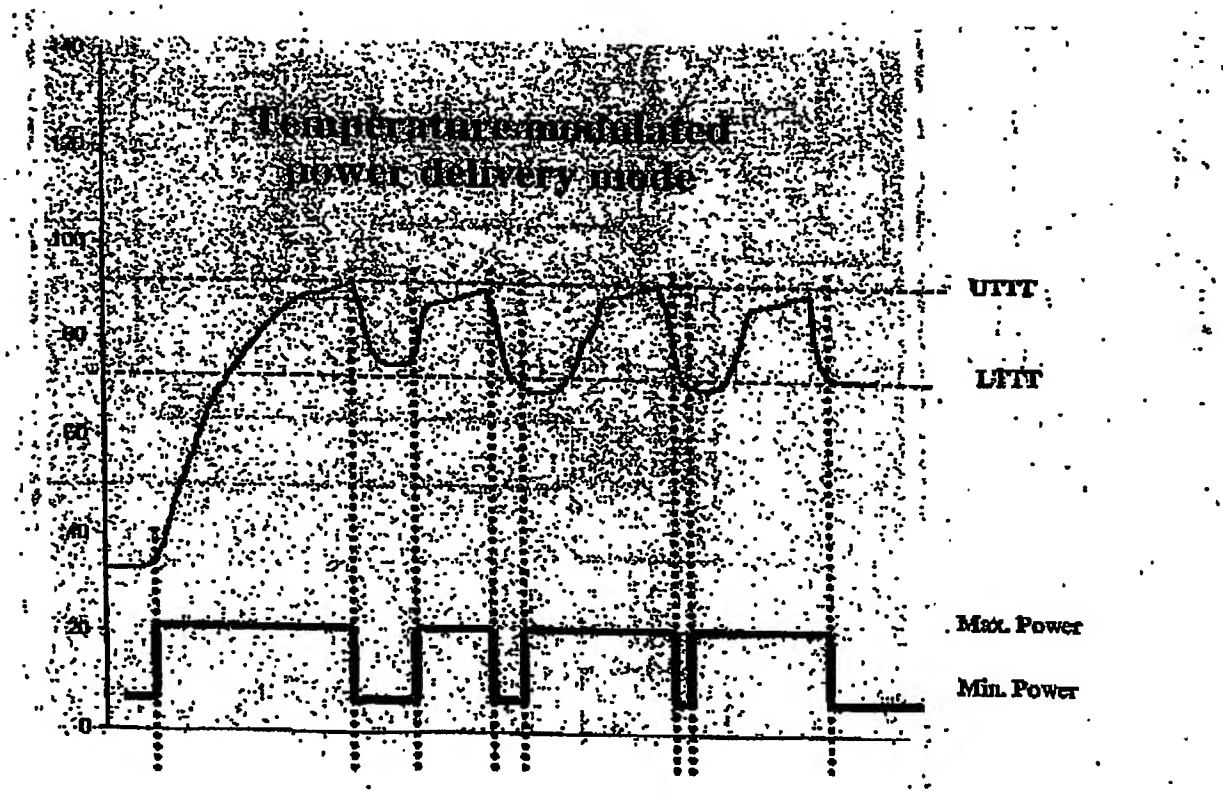
Temperature modulated
power delivery mode
UTTT
LTTT
Max. Power
Min. Power

Fig. 12

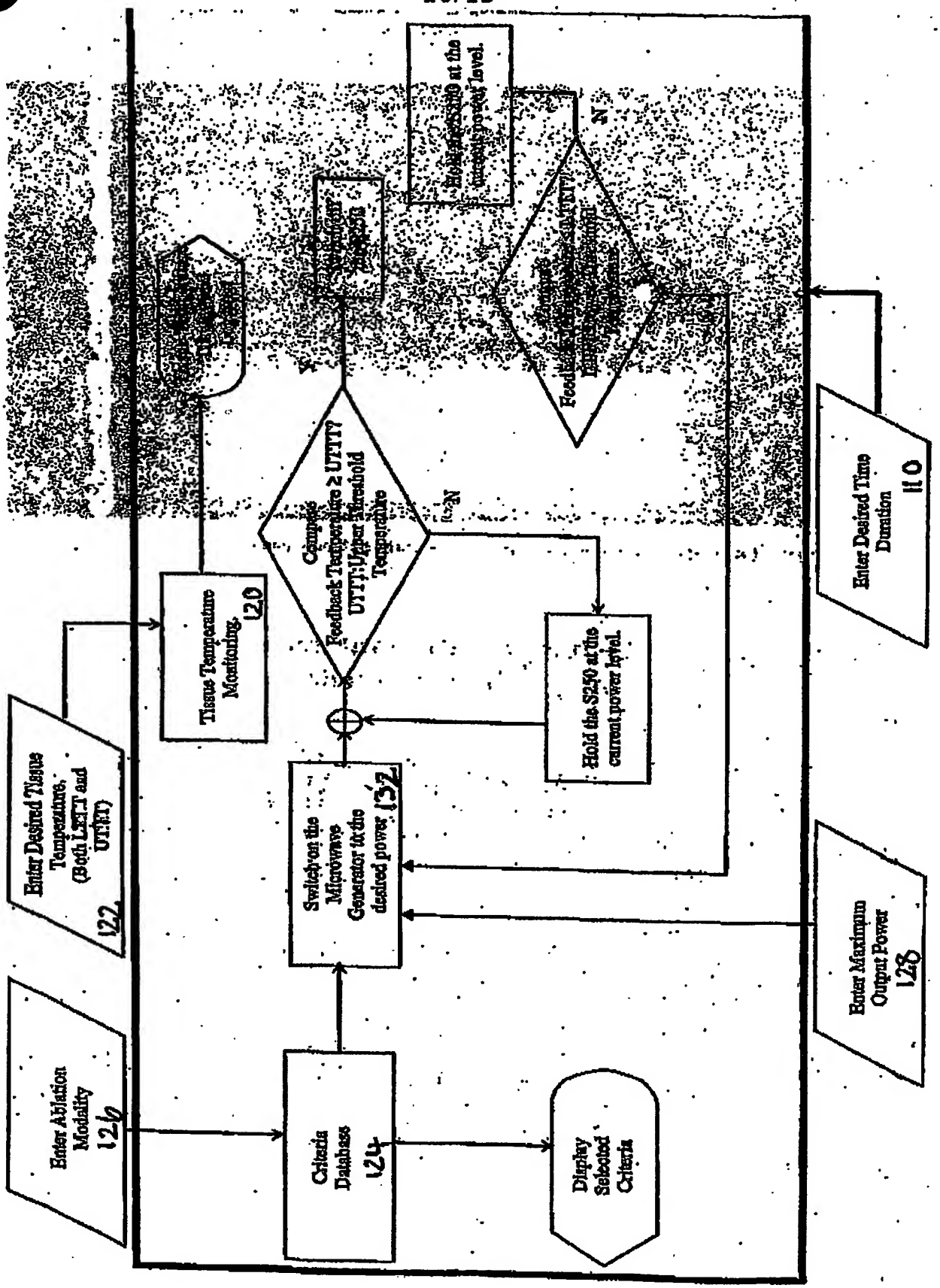
Enter Ablation Modality 126
Criteria Database 124
Display Selected Criteria
Enter Desired Tissue Temperature (Both LSTT and UTTT) 122
Tissue Temperature Monitoring 120
Switch on the Microwave Generator to the desired power 132
Hold the S250 at the current power level.
Compare Feedback Temperature ≥ UTTT/ Upper Threshold Temperature
Y
N
Feedback Temperature ≤ LSTT/ Lower Threshold Temperature
Y
N
Enter Desired Time Duration 110
Enter Maximum Output Power 128

Fig. 13

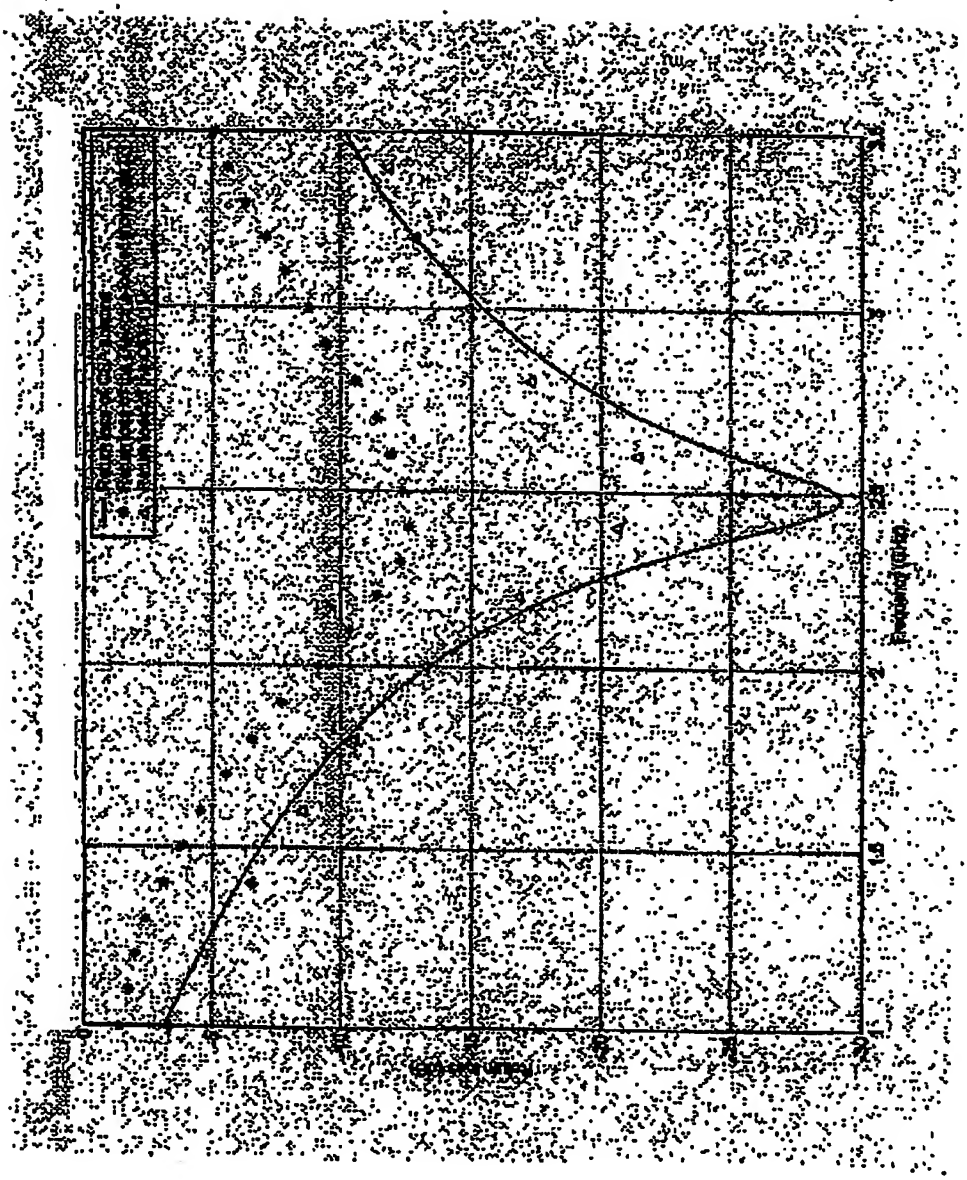

Fig. 14

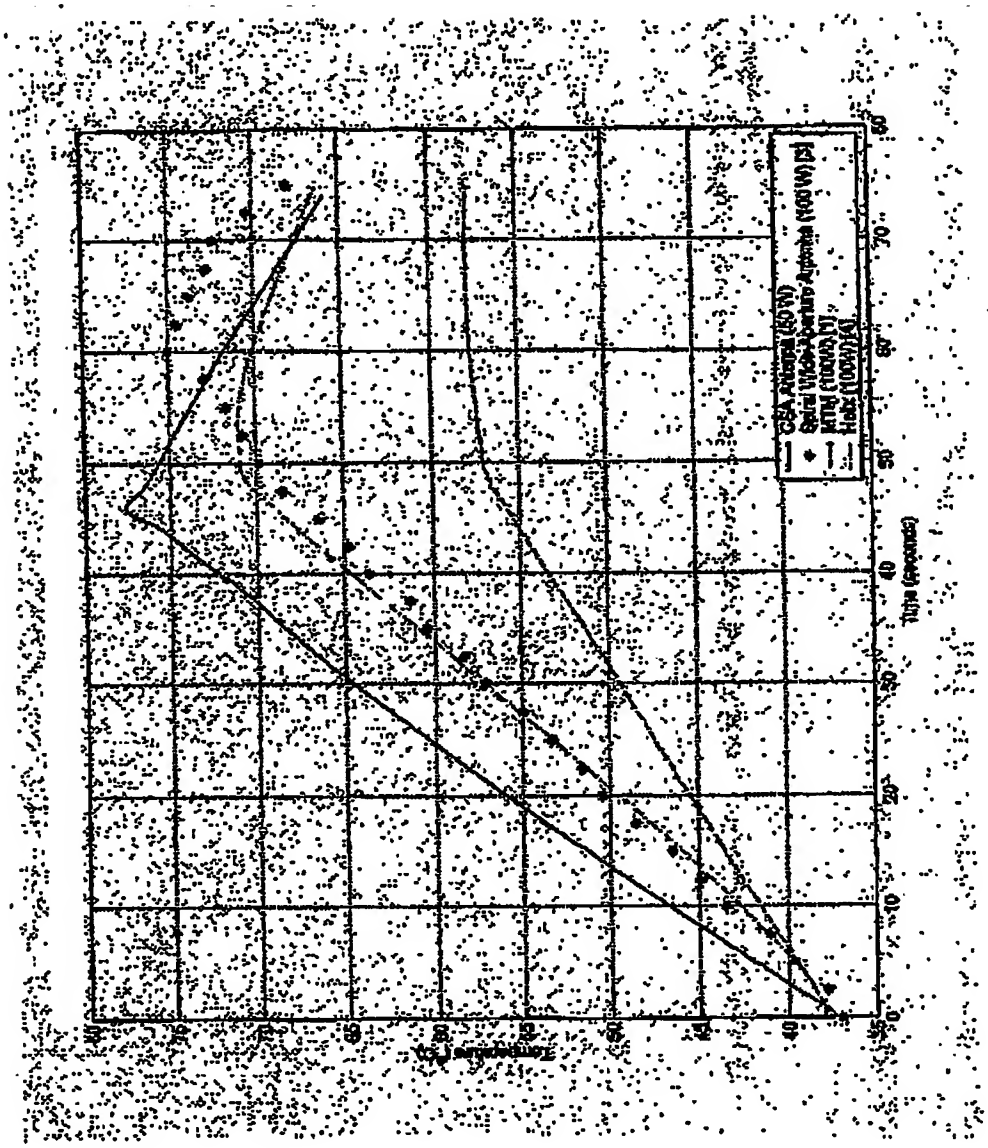

Fig. 15

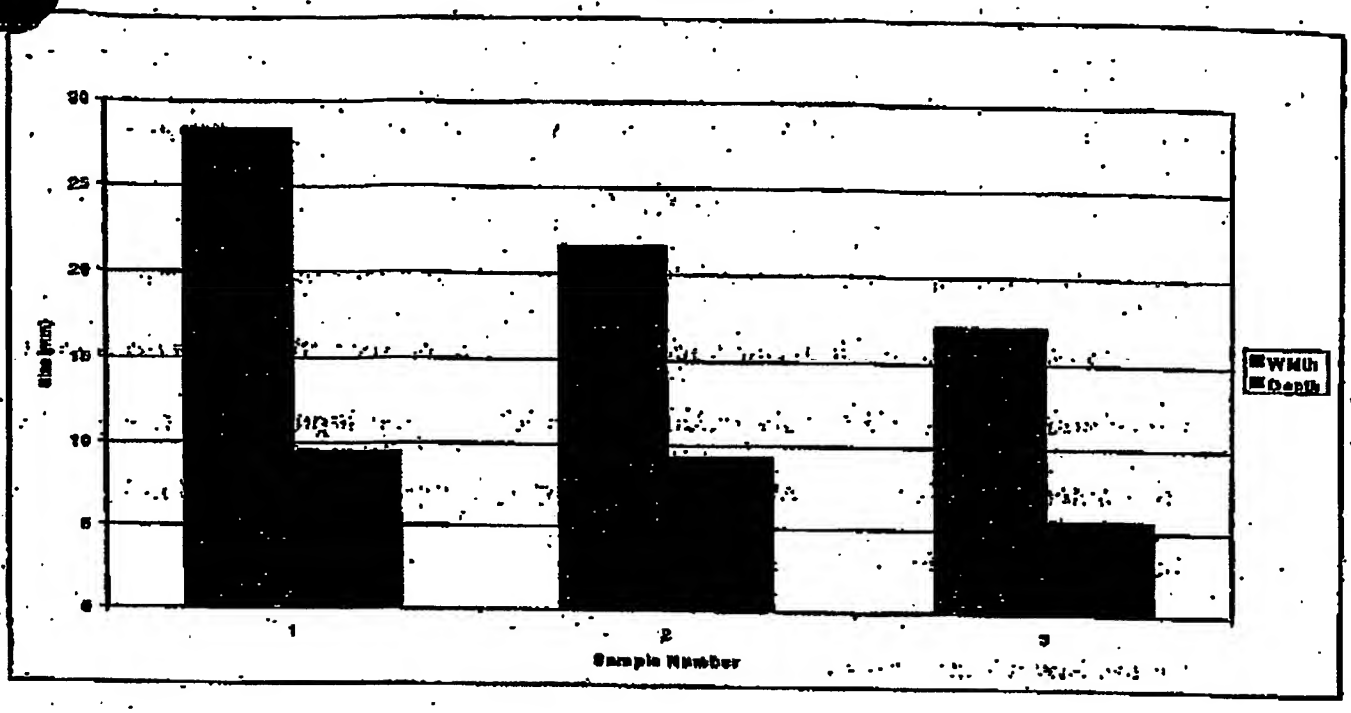
30
25
20
15
10
5
0
Size (mm)
1
2
3
Sample Number
Width
Depth

Fig. 16

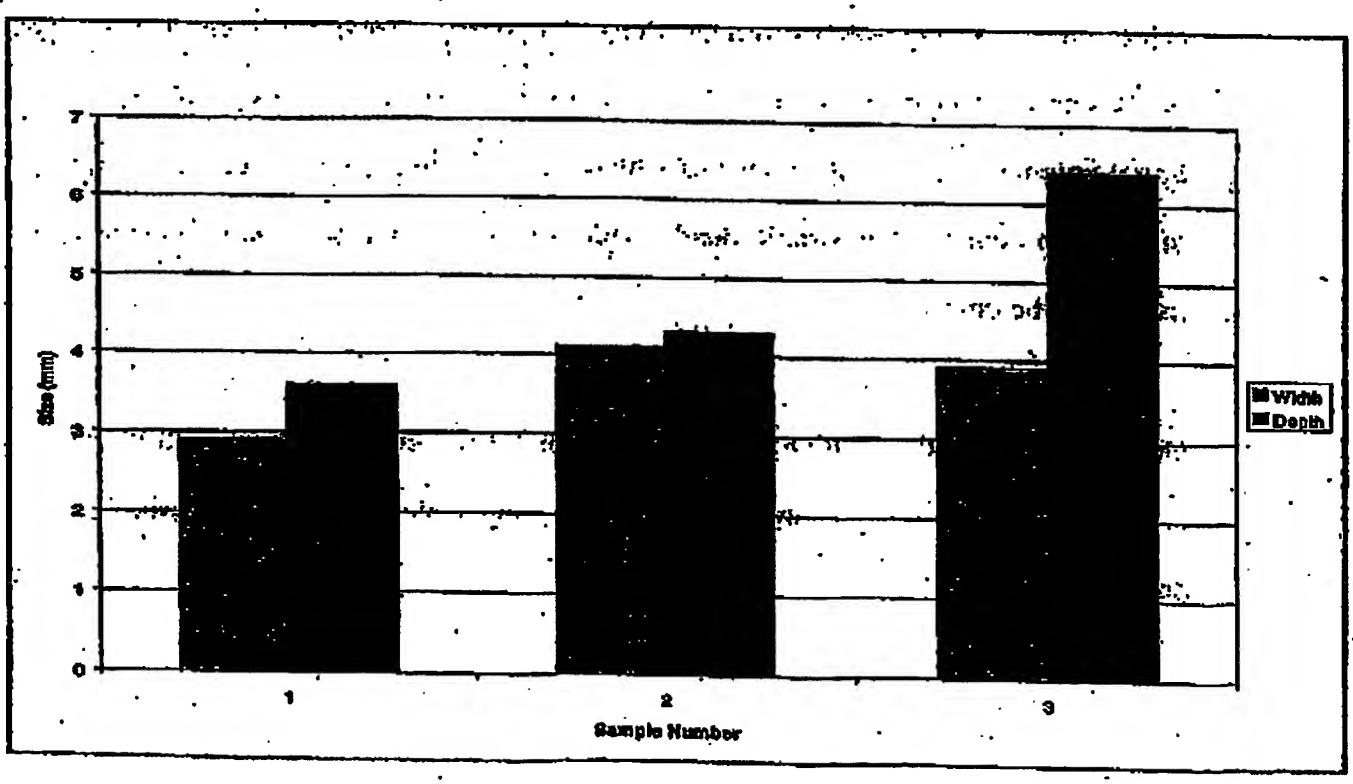
7
6
5
4
3
2
1
0
Size (mm)
1
2
3
Sample Number
Width
Depth

Fig. 17

Document made available under the Patent Cooperation Treaty (PCT)

International application number: PCT/AU04/000392

International filing date: 26 March 2004 (26.03.2004)

Document type: Certified copy of priority document

Document details: Country/Office: AU
Number: 2003901390
Filing date: 26 March 2003 (26.03.2003)

Date of receipt at the International Bureau: 22 February 2005 (22.02.2005)

Remark: Priority document submitted or transmitted to the International Bureau in compliance with Rule 17.1(a) or (b)